# BODY TO SENSOR CALIBRATION PROCEDURE FOR LOWER LIMB JOINT ANGLE ESTIMATION APPLIED TO IMU-BASED GAIT ANALYSIS

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**Abstract**: Inertial measurement units (IMU) are broadly used in gait analysis and have shown to be suitable to measure joint kinematics. The aim of this study is to investigate a calibration procedure to estimate body-to-sensor alignment in order to define anatomical quaternions. The procedure was performed in healthy subjects while remaining in standing position. After calibration, lower limb joint kinematics were assessed during gait trials. The results showed that the joint angles measured with this procedure were within the range of gait data reported in the reference literature, with interesting levels of low variation across trials (Table 2). In conclusion, the calibration procedure described in this work is simple, fast and do not require any additional tools or predefined/complex limb movements, which are desirable characteristics for use in clinical scenarios. Although the results were encouraging, further investigations on the reliability of such procedure will be performed in the near future.
**Keywords**: Gait analysis, inertial sensor, anatomical calibration, joint angular kinematics.

## Introduction

Walking performance is widely accepted as a general measure of functional ability [1]. For this reason, it is important to determine and assess kinematic and kinetic parameters, metabolic costs and/or muscle activity while walking. In persons with disabilities, this approach helps to decide the best treatment and to estimate progress during rehabilitation. Likewise, it is desirable that such evaluation take place in environments where patients perform their daily living activities, by means of a portable and easy to use device that is not restricted to laboratory settings.

Currently, wearable sensor systems offer real-time motion analysis without a complex set-up and are not limited to specific environments. Additionally, such devices present low weight and low cost, which make it suitable for ambulatory applications [2]. Sensors like accelerometers, gyroscopes and magnetometers are attached to different segments of the body to characterize and quantify human gait parameters. A multi-axis combination of gyroscopes, accelerometers and magnetometers is called Inertial Measurement Unit (IMU) [3]. This is a wide field of research, as clinical applications involving the use of IMU sensors are still largely unexplored in literature. The lack of standards of sensors placement on the body segments and the definition of its anatomical coordinate systems (CS) still limit the clinical application of this technology and further complicate the calculation of joint kinematics.

Different approaches to accomplish this task can be found in the literature. Luinge et al [4] proposed a coordinate system for upper arm and forearm according to predefined movements. During such movements, the direction of angular velocity determined one axis of the coordinate system. The other one was obtained by measuring the direction of the gravity vector and the last one by making an orthogonal system (cross product). An anatomical calibration technique using an external device was proposed by Picerno et al [5]. The device had two mobile pointers and one sensor aligned with the segment connecting them. The pointers were positioned at two anatomical landmarks that defined a body segment. For each segment, at least two non-parallel lines were determined to define the orthogonal frame by using a geometric rule.

Cutti et al [6] developed a protocol named "Outwalk" by defining many anatomical coordinate systems as the number of joints adjacent to the segment. The correct placement of the sensors (by identifying palpable landmarks) was required in this procedure, because the sensor frame attached to the pelvis was assumed to be coincident with the proximal pelvis coordinate. Additional functional movements needed to be performed in order to compute the knee flexion-extension axis.

The aim of this study is to investigate an auto calibration procedure performed before a walking trial in order to estimate lower limb joint kinematics. The procedure allows the estimation of body-to-sensor alignment by defining anatomical quaternions while the subjects remain in a standing, upright posture.

## Materials and methods

**Motion acquisition system** – The motion capture system Tech MCS (Technaid, Spain) was used in the experimental procedure. The device was connected via Bluetooth to a laptop and powered by AA batteries. In this study four Tech-IMU were used to obtain orientation data in real-time. Each IMU integrates three different types of sensors, an accelerometer, a gyroscope and a magnetometer, each one of three-dimensions. Data was acquired using Tech MCS Studio software in orientation quaternion format with a sampling frequency fixed to 50 Hz. MATLAB was used to analyze and process the orientation data. This commercial system presents errors smaller than 1° and was previously validated in [7].

**Kinematic model** - In the proposed kinematic model, the pelvis, thigh, shank and foot were assumed to be with uniform geometry and rigid segments [8]. Hip and knee kinematics are described by three angles [9]: flexion-extension, abduction-adduction and external-internal rotation. The sign of flexion-extension rotation for the knee is inverted to present positive values for flexion, as expected in clinical scenarios [6]. Ankle kinematics is also described by two angles [9]: dorsiflexion – plantar flexion and eversion-inversion. The pelvis quaternion was used as the anatomical quaternion (AQ) reference to define the thigh, shank and foot quaternions. Generally, a joint rotation is determined by measuring the orientation of a distal body segment with respect to a proximal segment [9]. We defined as many AQs as the number of joints adjacent to the segment, as proposed by Cutti et al [6]. For the pelvis and the foot, only one AQ was defined. For the thigh and the shank, two AQ were defined: thigh proximal AQ, thigh distal AQ, shank proximal AQ and shank distal AQ.

Figure 1. Sensor placement.

**Sensor placement** - Four sensors were positioned on pelvis and on right lower limb (thigh, shank and foot segments) (Figure 1). The pelvis sensor was placed on the sacrum at the S2 spinous process in the middle point between two posterior superior iliac spines. The IMU describes a coordinate system defined as X-axis pointing cranially and Z-axis pointing posteriorly. The thigh sensor was placed over the iliotibial tract approximately 5 cm above the patella. The shank sensor was positioned on the lower one-third of lateral shank 5 cm above of the lateral malleolus of the fibula.

The sensors on thigh and shank were positioned with X-axis pointing cranially and Z-axis pointing laterally. The foot sensor was fixed with double sided tape on the dorsal region of the foot over the 3rd and 4th metatarsal bones, 3 cm above to the corresponding metatarsophalangeal joints, with Z-axis pointing cranially and X-axis pointing posteriorly.

These sensors were attached with double-sided tape on an acrylic plate, which was glued to elastic band with Velcro. Such positions have been suggested by different authors [3], [6], [10].

**Calibration procedure** - During static acquisition (upright posture), the orientation data was used to define body-to-sensor alignment. The X-axis of the pelvis coordinate system of calibration was defined parallel to the vertical $Z_G$ (direction of gravity) and Z-axis, coincident with Z-axis of the sensor being aligned with the gravity.

Since the orientation data was obtained in quaternion format, the operations to correct or align the sensor ${}^{G}_{C}Q_{IMU-PELVIS}$ with the gravity were performed as follows:

1) Obtain X-axis ($X_{IMU-PELVIS}$) of coordinate system referred to the orientation of sensor measured by the quaternion associated ${}^{G}_{C}Q_{IMU-PELVIS}$.
2) Define the angle $\theta$ between X-axis and the gravity $Z_G$.
3) Define the vector $\boldsymbol{n_1}$ orthonormal to the mentioned vectors (X-axis and $Z_G$). Around this vector a rotation $\theta$ is made according to Euler's rotation theorem.

The angle $\theta$, the orthonormal vector $\mathbf{n_1}$ and the rotation $Q_{ROT}$ in quaternion representation were defined as:

$$\theta = cos^{-1}(X_{IMU-PELVIS} * Z_G) \tag{1}$$

$$\boldsymbol{n_1} = X_{IMU-PELVIS} \times Z_G \tag{2}$$

$$\boldsymbol{Q_{ROT}}(\theta, \boldsymbol{n_1}) = \left[\cos\left(\frac{\theta}{2}\right), \sin\left(\frac{\theta}{2}\right) \cdot \left[\frac{n_1}{\|n_1\|}\right]\right] \tag{3}$$

The pelvis quaternion (during calibration) with respect to the global system (G) was defined based on the sensor quaternion attached to the pelvis:

$${}^{G}_{C}Q_{PV} = Q_{ROT} \otimes {}^{G}_{C}Q_{IMU-PELVIS}, \tag{4}$$

where $\otimes$ denotes the product operator associated with quaternions and ${}^{G}_{C}Q_{PV}$ the pelvis quaternion (initial orientation of the pelvis and C refers to the calibration posture). Other initial anatomical quaternions were defined during the calibration procedure as shown in Table 1.

Once the anatomical quaternions were defined, body-to-sensor orientation ${}^{B}Q_{IMU-B}$ was determined for each sensor:

$${}^{B}Q_{IMU-B} = {}^{G}_{C}Q^{*}_{B} \otimes {}^{G}_{C}Q_{IMU-B} \tag{5}$$

where B denotes the body segment and $*$ the complex conjugate of the quaternion. Having the initial relative orientation of the sensor to body segment, the orientation of each segment at any instant in time can be determined. Then, the joint rotation is defined as:

$$Q_{JOINT} = {}^{G}Q^{*}_{pB} \otimes {}^{G}Q_{dB} \tag{6}$$

Table 1: Definition of anatomical quaternions obtained during calibration posture (straight upright posture)

| Segment | Initial quaternion definition |
|---|---|
| Pelvis (PV) | ${}^{G}_{C}Q_{PV}$ |
| Thigh proximal (pTH) | ${}^{G}_{C}Q_{pTH} = {}^{G}_{C}Q_{PV}$ |
| Thigh distal (dTH) | ${}^{G}_{C}Q_{dTH} = {}^{G}_{C}Q_{PV} \otimes Q_{ROT}(90°, \boldsymbol{x})$ |
| Shank proximal (pSH) | ${}^{G}_{C}Q_{pSH} = {}^{G}_{C}Q_{dTH}$ |
| Shank distal (dSH) | ${}^{G}_{C}Q_{dSH} = {}^{G}_{C}Q_{pSH} \otimes Q_{ROT}(180°, \boldsymbol{n_2})$ |
| Foot (FT) | ${}^{G}_{C}Q_{FT} = {}^{G}_{C}Q_{dSH}$ |

Where $\boldsymbol{x} = [1\ 0\ 0]$ and $\boldsymbol{n_2} = [1\ 0\ 1]$ around which the rotation occurs. ${}^{G}_{C}Q_{B}$ the relative orientation of body segment (during calibration posture) AQ with respect to the global system. $Q_{ROT}(\theta, n)$ the rotation $\theta$ about $n$ according to Euler's rotation theorem.

where $pB$ denotes the proximal and the $dB$ the distal segment respect to the global system. The rotation $Q_{JOINT}$ describes joint angles, which can be extracted using the Euler sequence ZXY [6].

**Experimental protocol –** Five volunteers without gait disabilities (3 men and 2 women, 26 ± 4 years old) were enrolled in this study. All sensors were placed in the aforementioned sites by a trained physiotherapist. The sensor placed to the pelvis was aligned to the walking direction. The subjects were asked to keep a straight, upright posture during 5 seconds before start walking in a 10 meters walkway. This calibration posture allowed the definition of the body-to-sensor alignment. Each subject performed three trials and the three middle gait cycles were extracted for further analysis. A total of nine gait cycles were acquired for each subject. This research was approved by the Ethical Committee of UFES (research project 214/10).

**Kinematic parameters –** The variables evaluated in this study were the discrete angular kinematic parameters previously reported in a reference work by Benedetti [9] for the three planes of motion.

In the sagittal plane, K1, H1 and A1 refer to flexion at heel strike for knee, hip and ankle, respectively. K2, H2 and A2: peak flexion (plantar flexion for ankle) at loading response. K3, H3 and A3: peak extension (dorsiflexion for ankle) in stance phase. K5, H5 and A5: peak flexion (dorsiflexion for ankle) in swing phase.

Frontal plane variables were K8, H8 and A8, which denote peak adduction (eversion for ankle) in stance phase. K9, H9 and A9 refer to the peak abduction (ankle inversion) in swing. Transverse plane was assessed by the pairs K11, H11 (peak internal rotation in stance) and K12, H12 (peak external rotation in swing phase).

## Results

As a representative case, we will present data of one subject (subject #2). Figure 2 shows the mean and standard deviations of the angular displacement for the lower limb joints (in stride percentage), calculated using the body-to-sensor calibration procedure proposed in this work. Table 2 show the parameters obtained by the proposed algorithm and the reference work by Benedetti [9].

Similar results were observed for all 5 participants. Knee Ab-Adduction and Int-External Rotation plots are reported over a gray background because these rotations are not reliable due to soft-tissue artifact [7], [11]. The largest standard deviation (5.2°) was observed in ankle flex-extension.

## Discussion

This work investigated a novel calibration procedure for IMU-based gait analysis performed in a static position before the walking trial. The proposed strategy was designed for simplicity, accuracy and ease-of-use in a clinical setting.

The obtained joint angles (mean and standard deviation) were consistent with those presented in the reference literature [9], including both signal behavior and the intervals at which it occurs [5, 7, 8, 9, 10].

The largest amplitude occurred on the sagittal plane (flexion-extension) as expected. The knee motion on the frontal plane was not consistent with the findings of Benedetti [9]. However, previous research has shown that movements in this plane have a broader range of variation, which includes the interval of our results [7, 8].

Interestingly, the results obtained with the developed algorithm presented low standard deviations (Table 2), which means that estimated measures were consistent across trials. Further investigation on the reliability of this procedure and also internal/external validity of the results will be performed in the future.

## Conclusion

This study implemented a novel, simple and fast calibration procedure to address the problem of body-to-sensor alignment in IMU-based gait analysis, providing tridimensional kinematics of hip, knee and ankle with only 4 IMUs, without resorting to any additional tools or predefined movements. This procedure has the potential to become an alternative to camera-based system in applications in which patients perform their daily activities.

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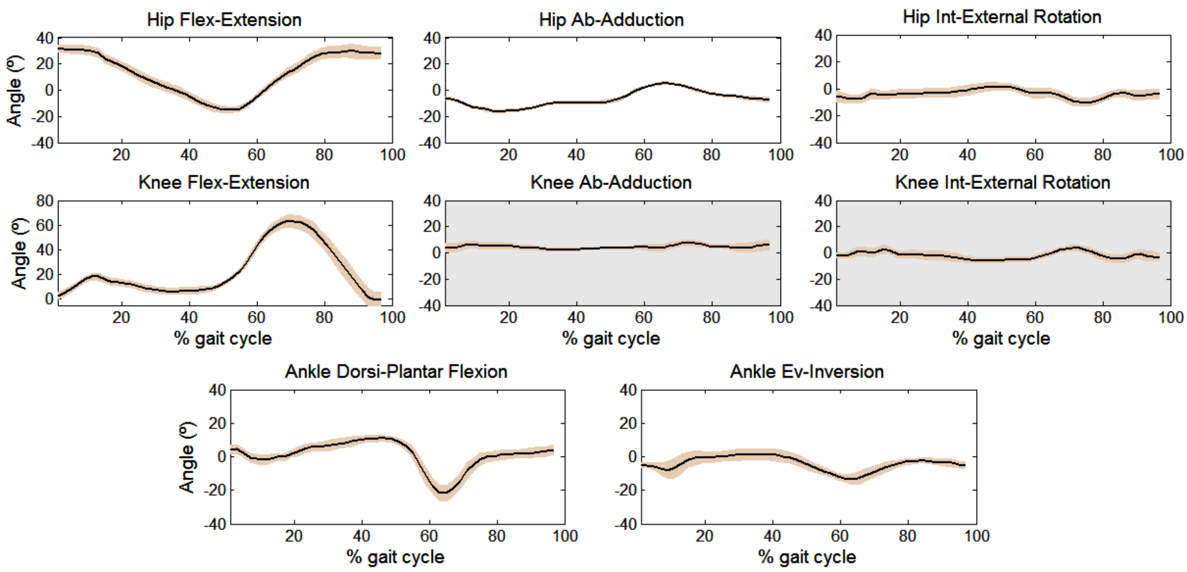


Figure 2. Joint angular kinematics in stride percentage (from heel strike to heel strike) of one subject. Nine gait cycles were summarized by black curve (mean) and orange stripe (± std).

Table 2. Mean (standard deviation) of the gait analysis parameters

| | | Flex-Extension (º) | | | | Ab-Adduction (º) | | Int-External Rot (º) | |
|---|---|---|---|---|---|---|---|---|---|
| | | **H1(std)** | **H2(std)** | **H3(std)** | **H5(std)** | **H8(std)** | **H9(std)** | **H11(std)** | **H12(std)** |
| **HIP** | Algorithm | 31.3 (3.2) | 29.9 (3.9) | -15.0 (2.8) | 29.4 (4.7) | -16.3 (1.5) | 5.2 (1.4) | 1.5 (3.0) | -10.4 (3.3) |
| | Benedetti [9] | 26.7 (5.4) | 28.9 (5.7) | -10.0 (5.1) | 29.8 (4.8) | -5.4 (3.3) | 5.4 (3.3) | 3.42 (4.9) | -8.5 (6.0) |
| | | **K1(std)** | **K2(std)** | **K3(std)** | **K5(std)** | **K8(std)** | **K9(std)** | **K11(std)** | **K12(std)** |
| **KNEE** | Algorithm | 2.9 (3.1) | 18.1 (3.0) | 6.1 (3.1) | 63.3 (5.1) | 2.2 (1.7) | 7.1 (2.8) | -3.0 (3.7) | 4.0 (2.4) |
| | Benedetti [9] | 0.39 (4.9) | 17.9 (7.8) | 4.9 (4.6) | 65.6 (5.2) | 3.1 (3.6) | -4.0 (10.4) | 5.2 (5.3) | -8.4 (5.8) |
| | | **A1(std)** | **A2(std)** | **A3(std)** | **A5(std)** | **A8(std)** | **A9(std)** | | |
| **ANKLE** | Algorithm | 3.9 (3.0) | -2.0 (3.1) | 10.9 (2.2) | -21.9 (5.2) | 1.3 (3.3) | -13.3 (3.5) | | |
| | Benedetti [9] | -4.0 (6.0) | -12.7 (5.0) | 10.9 (5.7) | -22.6 (7.0) | 3.2 (4.0) | -9.2 (4.4) | | |

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